In Vivo Human Coronary Magnetic Resonance Angiography at 7 Tesla

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Introduction. Cardiac MRI at high fields faces many challenges including the lack of commercially available body RF coils, increased sample-induced B1 inhomogeneity, increased magnetic susceptibility effects which make imaging with balanced sequences difficult, and SAR limitations. However, if these can be at least partially overcome, then the higher signal-to-noise is advantageous both for imaging and localized spectroscopy. Here we investigate the feasibility of acquiring coronary magnetic resonance angiography (CMRA) scans, a promising technique for the non-invasive visualization of the coronary anatomy (1), in volunteers at 7 tesla (T) on a time-scale acceptable for clinical studies.

Methods. Eleven healthy volunteers (8 men, 3 women, mean age 33.6 ± 8.8 years) were imaged in a 7T MR system (Philips Healthcare, Best, NL). Since no body coil is available, a multi-segmented 13-cm diameter anterior surface coil was designed and constructed and used in transmit/receive mode. A commercial vector ECG (VECG) system was used for R-wave triggering. Volume selective RF power optimization and shimming were applied for each scan. A segmented k-space gradient echo sequence was used for scout scanning. Multi-slice cine scans were used for coronary artery localization and for the visual identification of the time period (Td) of minimal coronary motion. Scan plane localization parallel to the right coronary artery (RCA) was facilitated using a three-point planscan tool. Double-oblique free-breathing 3D coronary MRA (segmented k-space gradient-echo imaging, TR=4ms, TE=1.5ms, RF excitation angle=15°, field-of-view=420x269x30mm^2, scan matrix=512x488, 30 slices, slice thickness=2mm, acquisition window~100ms, scan time~4min) was performed using prospective navigator gating with the 2D selective navigator localized at the left heart-lung interface. Image data were collected in mid-diastole at the predetermined Td. An adiabatic spectrally selective inversion recovery pre-pulse (TI=200ms) was used for fat suppression and enhanced contrast between the coronary blood-pool and epicardial fat. Coronary MRAs were reformatted and measurements were performed using the "Soapbubble" software tool (2).

Results. Right coronary MRAs were successfully obtained in all eleven healthy adult human subjects. The mean duration of the 3D coronary MRA scan was 240 ± 32 seconds. The average measured contiguous length of the RCA was 6.8 ± 3.6 cm. The average measured diameter of the first 4 cm of the RCA was 3.0 ± 0.6 mm. The signal-to-noise of the blood-pool was measured to be 18.5 ± 13.4, with an average vessel sharpness of 43 ± 10%.

Figure 1

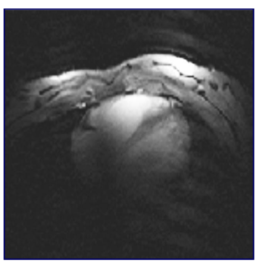

Figure 2

Figure 3

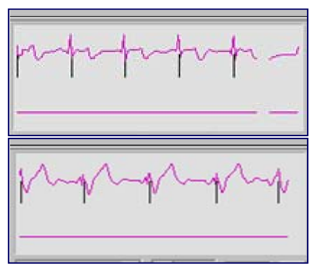

Figure 4

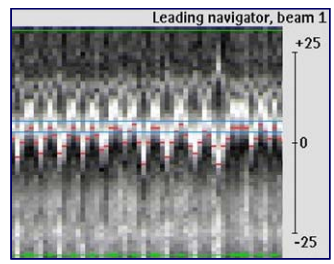

Figure 5

Figures 1 and 2 (left) show scout images in the coronal and axial plane, respectively, illustrating that the RF penetration of the 13 cm diameter surface coil is sufficient for RCA imaging. Figure 3 (centre) shows a long contiguous segment of the RCA (arrow) with high contrast between the blood-pool and the epicardial fat. One potential concern is that, at the higher field strength, the magneto-hydrodynamic effect is amplified with an artificial augmentation of the T-wave of the ECG (upper Figure 4 shows an ECG outside the MR scanner, lowest Figure 4 an ECG inside the 7T MRI). Nevertheless, the VECG algorithm allowed reliable R-wave triggering. In Figure 5 (right) the navigator signal from the heart-lung interface received by the surface coil can be seen.

Discussion. This work represents the first report of human coronary MRA at 7T using appropriate adaptations of the scanning protocol at lower fields and the use of a custom-built transmit-receive surface coil. Long contiguous segments of the RCA with reasonable contrast and acceptable scanning times can be obtained. Future work will focus on optimizing contrast enhancement between the blood-pool and the myocardium within the SAR and B1 homogeneity constraints. In order to improve volumetric coverage, the development of larger surface coils or coil arrays will be required.

References.

1. Stuber M et al., J of Magnetic Resonance Imaging 2007, 26, 219-234
2. Etienne A et al., Magnetic Resonance in Medicine 2002, 48, 658-666